Coincident Compton Nuclear Medical Imager

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***Abstract*-- A novel method is proposed for nuclear medical imaging. A high resolution Compton imager with large solid angle and high efficiency is used to detect multiple gamma rays from selected radionuclide decays. The direction cone of each gamma ray is re-constructed from the multiple Compton interactions in the detector array. The intersection of three Compton direction cones defines a limited number of positions, often a unique location within the region of interest, for each decay. For positron decay radionuclides that are accompanied by a coincident gamma ray (e.g. ^{14}O) the location is determined from the intersection of the 511 keV interaction sites and the Compton direction cone of the third gamma ray. This technique has the advantages of 3-dimensional imaging with lower doses, and comparable or improved resolution compared to standard SPECT and PET techniques. The technique will be described, along with a discussion of alternative detectors configurations. Results of simulations showing the capability of this technique are presented.**

I. Introduction

The two most widely used nuclear medical imaging techniques are PET and SPECT. For each nuclear decay, SPECT determines the line of origin using a collimated, position-sensitive detector. A two-dimensional image is obtained from a single view angle. Three-dimensional images can be re-constructed using gamma ray views from many different directions. Advantages of SPECT are the wide range of radiopharmaceuticals available and the lower cost relative to PET. A significant disadvantage is the loss of most of the emitted gamma rays in the collimator (typically only one in 10^3 to 10^4 are detected). PET determines a line of origin for the nuclear decay from the interaction sites of two 511 keV gamma rays. An advantage of PET is that no collimation is required, and somewhat better imaging resolution (4-6 mm for PET vs. 7-10 mm typical SPECT systems), however the costs of PET systems are higher.

Several groups have considered Compton medical imagers as an alternative to conventional SPECT imaging [1]-[7]. In Compton imaging, the initial gamma ray emitted by the radionuclide undergoes a Compton scatter interaction in a first position-sensitive detector, and the scattered gamma ray undergoes full energy absorption through the photoelectric effect in a second position-sensitive detector. With the knowledge of the energy losses and positions of the two interactions, the angle of scatter is given by the Compton scatter formula:

$$\cos J = 1 - m_e c^2 \left(\frac{1}{E_1} - \frac{1}{E_0} \right) \qquad (1)$$

where E_0 is the incident gamma ray energy, E_1 is the scattered gamma ray energy, $m_e c^2$ is the rest mass of the electron, and q is the scatter angle. The direction of the gamma ray is then restricted to a conical surface with apex at the location of the first interaction site, axis along the line connecting the two interaction sites, and with a half opening angle given by the Compton scatter angle. Backprojecting the direction cones for a large number of gamma rays can be used to re-construct a three-dimension image of the radioactivity.

Eliminating the collimator in Compton "electronically collimated" devices is attractive. In principle, it can enable a much larger fraction of the emitted gamma rays to be used, relative to SPECT, in the image reconstruction. Furthermore, Compton imaging could be effective at energies above several hundred keV where SPECT is not very suitable due to the difficulty of constructing effective collimators. The potential for improved imaging resolution and efficiencies relative to conventional SPECT make this technique attractive. However, the low efficiency of the small Compton devices considered up to this time is a serious drawback.

A further consideration with Compton devices is the Doppler broadening which limits the angular resolution that can be achieved, especially at low gamma ray energies. The standard Compton formula assumes the scattering takes place with a free electron at rest. Taking into account the pre-interaction momenta of orbital electrons introduces uncertainties in the energy and angle of the scattered gamma ray [8],[9].This is reflected in an uncertainty in the direction of the incoming gamma ray. This uncertainty is smaller for higher energy gamma rays and for lower-Z materials [10]. For 511 keV gamma rays undergoing Compton scatter in germanium, the uncertainty is typically about 1 degree.

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II. Coincident Compton Imaging

A significant improvement relative to single photon Compton imaging can be realized, in principle, with coincident Compton imaging. Improved imaging with reduced dose could be achieved if the location for each decay could be determined uniquely. This is possible with the use of selected radio-nuclides that emit a three-gamma cascade and positron-decay isotopes that are accompanied by a coincident gamma ray. ^{94}Tc and ^{14}O are respective examples. Liang et al. [11] proposed this technique, and compared the relative capabilities of SPECT and their triple gamma coincidence tomographic imaging concept. Hart [12] was granted a patent for this concept.

Consider first the case of a positron-decay nuclide with the emission of a coincident gamma ray as shown in Fig. 1. ^{14}O is an example. The two 511 keV gamma rays will interact in the position-sensitive detector array in a standard PET mode. The locations of the 511 keV interactions define a pencil beam along which the decay occurred. The third gamma ray Compton scatters in the detector array at a first interaction site, and the scattered gamma ray interacts by Compton scattering or photoelectric effect at a second interaction site. A Compton direction cone for the latter event will intersect the PET pencil beam at one, or perhaps two locations in the region of interest. Thus, the origin of the decay is determined to a point rather than a line. There are a large number of radio-isotopes that can be utilized, including ^{14}C, ^{42m}Sc, ^{44}Sc, ^{48}V, ^{50}Mn, ^{52m}Mn, ^{54}Co, ^{55}Co, ^{60}Cu, ^{66}Ga, ^{67}Ge, ^{70}As, and ^{73}Se.

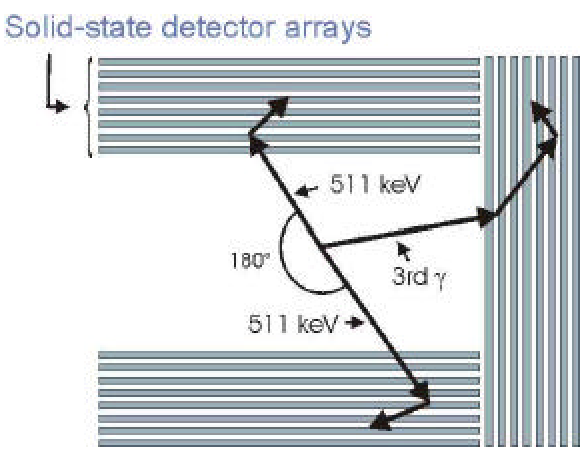

Fig. 1. Coincident Compton imager using a positron-emitting radionuclide accompanied by an additional gamma ray.

Similarly, for three coincident gamma rays emitted in a cascade, each of the gamma rays can be restricted to a direction cone. The intersection of the three direction cones defines a limited number of regions where the decay may have occurred—typically only one in the region of interest. Radio-isotopes that have decays with a cascade of three of more gamma rays include: ^{28}Mg, ^{48}Sc, ^{71m}Zn, ^{81}Br, ^{94}Tc, ^{94m}Tc, and ^{131}I. ^{94}Tc appears to be a particularly attractive choice due to its three high energy gamma rays at 702, 850 and 871 that are emitted in 100% of the decays, the low Doppler broadening associated with these high energy gamma rays, the half life of 6 hrs., and the wide range of radio-pharmaceuticals that have been developed for ^{99m}Tc.

The angular resolution of the Compton imaging technique

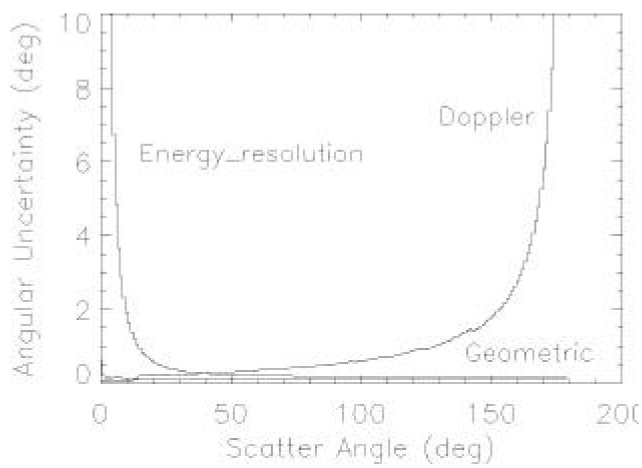

Fig. 2. Angular uncertainties in Compton scattering due to Doppler broadening, detector energy resolution (2 keV) and detector position resolution (0.5 mm) for 750 keV incident gamma ray scattering in silicon.

depends on the position and energy resolutions in the detectors and the Doppler broadening effect. Ordonez et al. have investigated these effects in some detail [8], [9]. The ultimate limitation is imposed by Doppler broadening, and this is optimized by using higher-energy gamma rays and lower-Z detectors. Use of germanium and/or silicon position-sensitive detectors is preferred. Momayezi et al. [13], Amman et al. [14] and Wulf et al. [15] have demonstrated 3-D position sensitivity in germanium strip detectors. Figure 2 shows the angular resolution that can be achieved using 750 keV gamma rays (typical of the ^{94}Tc lines) as a function of Compton scatter angle. The locations of interactions were assumed to be determined to 0.5 mm, and the energy resolution was assumed to be 2 keV FWHM. The angular resolution is seen to be 0.5 – 1 degree for Compton scatter angles between 20° and 120°. This corresponds to 2 mm resolution at a distance of 20 cm. With this resolution, three-dimensional imaging competitive with or better than clinical PET should be achievable.

Advantages of the triple coincident Compton imaging include: competitive imaging resolution and lower patient dose. Disadvantages include: need for expensive position sensitive detectors, reduced efficiency due to the requirement for triple coincidence, and the need for large solid angle detector arrays.

Liang et al. [11] presented the concept for triple coincident Compton imaging. However, even though it has several significant advantages, this technique has not found research or clinical applications. This is due in large part to the limited availability of the 2-D position-sensitive germanium detectors that were proposed. Several developments make consideration of this technique more attractive at this time. These include the development of 3-D position-sensitive germanium detectors, the development of position-sensitive silicon detectors that will reduce the Doppler broadening effect, the use of multiple Compton scattering wherein the direction and energy of the incident gamma rays can be determined without requiring full energy deposition (thereby providing increased efficiencies) [16], [17] and the development of low-power, spectroscopy application specific integrated circuits (ASICs).

One approach, albeit an expensive approach, is to utilize large arrays of position-sensitive solid-state detectors (Ge, Si, CZT) as shown in Fig. 1 and Fig. 3. With the large solid angle and thick detectors, the efficiency of this approach can be relatively high.

Consider the efficiency of the system in Fig. 3 for Compton imaging with ^{94}Tc using the 702, 849 and 871 gamma ray cascade. Following Hart [12], we estimate the efficiency for reconstruction of each gamma ray in two applications: small animal imaging and a patient head scan. The efficiency is the combined probability of several factors:

$$Eff = P_{esc} * P_{\Omega} * P_{1st} * P_{2nd} * P_{20\text{-}120}$$

where:

P_{esc} = probability for 750 keV gamma ray escaping the subject without interaction (e.g. 10 g/cm^2 typical of a head scan)
P_{Ω} = solid angle subtended by detector array ($\geq 3\pi$ ster.)
P_{1st} = probability for Compton scatter in array at first site (we assume 25 g/cm2 detector thickness)
P_{2nd} = probability for interaction in array at second site
$P_{20\text{-}120}$ = probability for scattering between 20° and 120°.

Table 1 lists the estimated probabilities along with the combined probability for detection of a single gamma ray, P_{single}. The probability for a single ^{94}Tc gamma ray being detected in a head scan device is about 0.14. The combined probability for detecting the three ^{94}Tc gamma rays (702, 850 and 871 keV), P_{triple}, and having the three be at angles greater then 20 degrees from each other (good imaging events) is ~0.001. Therefore, if a 10^6 Bq (27 micro-curies) source is administered, the triple coincident rate will be ~1000 counts/s. In a 1000 s duration exam, 10^6 decays would be resolved in 1 mm - 2 mm voxels, enabling a high quality 3-dimensional image to be constructed.

Table 1. Triple coincidence detection probabilities

	P_{esc}	P_{Ω}	P_{1st}	P_{2nd}	$P_{20\text{-}120}$	P_{single}	P_{triple}
Small Animal							
^{94}Tc	0.82	0.9	0.82	0.7	0.75	0.32	0.030
^{14}O							0.10
511 keV	0.78	0.9	0.87			0.61	
2.31 keV	0.89	0.9	0.65	0.7	0.75	0.27	
Head							
^{94}Tc	0.43	0.75	0.82	0.7	0.75	0.139	0.0027
^{14}O							0.0079
511 keV	0.35	0.75	0.87			0.23	
2.31 keV	0.6	0.75	0.65	0.7	0.74	0.15	

III. Implementation Options

A. Full Arrays with 3-D Position-sensitive Detectors.

Shown in Fig. 3 is an "ideal" detector configuration. The detector consists of many layers of 3-D position-sensitive detectors. A large solid angle, greater than 3π ster, and multiple layers of detectors, are used to maximize the probability for detection of each of the gamma rays. Use of low-Z detectors (Si, Ge) is preferred to ensure high probability for the first event to be a Compton-scatter interaction. Ideally the second interaction would be a photoelectric event, but this requires a high-Z detector. Full energy absorption for each gamma ray would enable clear energy identification of each incident gamma ray, providing

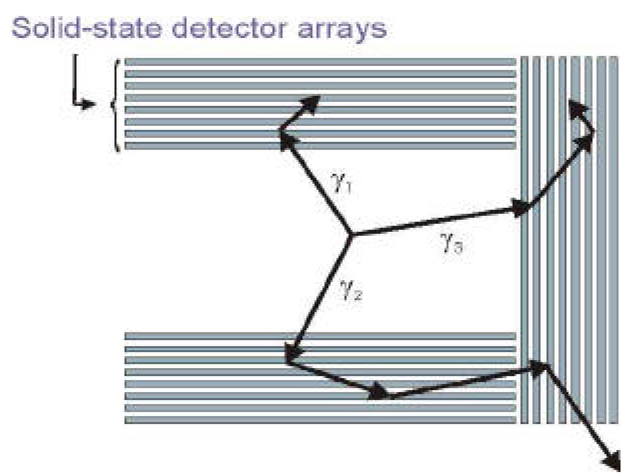

Fig. 3. Coincident Compton imager using 3-gamma cascade.

unambiguous position from the intersection of the three Compton direction cones.

Consider the case where the full energy of each initial gamma ray is not deposited. The energy of the initial gamma ray can still be determined if there are two Compton scatters followed by a third interaction, the energy losses at the first two sites are determined and the position of the three sites are determined [16], [17]. In this case the energy resolution of the initial gamma rays are less well determined due to the

Doppler broadening at the first interaction site. However, the energy resolution should still be adequate to clearly separate the three ^{94}Tc gamma rays [18].

Now suppose that only two interactions occur before the scattered gamma ray exits the detector system. In this case, it may still be possible to determine the most probable energy if the other two gamma rays are determined.

B. Alternative Detector Configurations

Figures 1 and 3 show a detector configuration with many layers of position-sensitive solid-state detectors. This is the preferred approach for the energy of each gamma ray can be determined accurately, thereby providing the best imaging resolution. This is also an expensive implementation.

There are several alternative implementations that can be considered that would be less expensive in detector technology, but that may still provide good imaging. The scatter and absorber detectors can be functionally separate. For example, the scatterer could be an array of position-sensitive solid-state detectors and the absorber could be an array of position-sensitive scintillation detectors that surround the scatterer array. The several interactions in each detector array would be analyzed to determine the most probable sequence of interactions for each gamma ray. With the modest energy resolution of the scintillation detectors, this could result in misidentifications for the proper gamma ray energies, e.g. in the case of the ^{94}Tc 850 and 872 keV lines. In this case, two sets of Compton direction cones would be generated: one giving the correct location for the decay and on that would be offset slightly. The resulting image formed using both combinations would give a true image of the radioactivity with an additional halo added to it. Simulations will be undertaken to further study how serious this effect would be.

Another configuration could have the scatterer be a position-sensitive detector with poor energy resolution and the absorber be an array of solid-state detectors. Again, the alternative energies for the coincident gamma rays would be considered, and the energy loss at the first scatter site would be taken to be the difference between the assumed initial gamma ray and the energy loss in the solid-state array. This approach is similar to that proposed by Rohe and Valentine [5] for single photon Compton imaging.

IV. Simulations

A Monte Carlo simulation has been run to show the capability for imaging a 3-dimensional object using the coincident Compton image technique. The simulation assumed ^{94}Tc uniformly distributed in five cubic volumes: a 4mm x 4mm x 4mm central cube, with four 2mm x 2mm x2mm cubes located on four of the eight corners of the central cube. ^{94}Tc coincident gamma rays were detected by a first detector array located at distances of 5 cm, 10 cm, 20 cm, or 30 cm from the center of the distribution. The scattered gamma ray was detected by a second detector array separated by about 10 cm from the first detector. The central region encompassing the radioactive regions was divided into1 mm^3 voxels, and each voxel was tested for consistency with the conical shell for each of the three gamma rays. Those voxels consistent with all three Compton cones are treated as a valid location for the radioactive decay. The uncertainty in the Compton scatter angle included Doppler broadening at the first interaction site, and the energy and position resolutions of the detectors. This process was repeated for a significant number of radioactive decays, producing a summed distribution of possible source regions. Surface maps of the regions are shown in Fig. 4 for source to first detector distances of 5 cm, 10 cm, 20 cm and 30 cm. It is seen that, with this simplified simulation, an image resolution of 2 mm is realized for a source to first interaction distance of up to 20 cm, consistent with that expected based on the limitations of Doppler broadening at

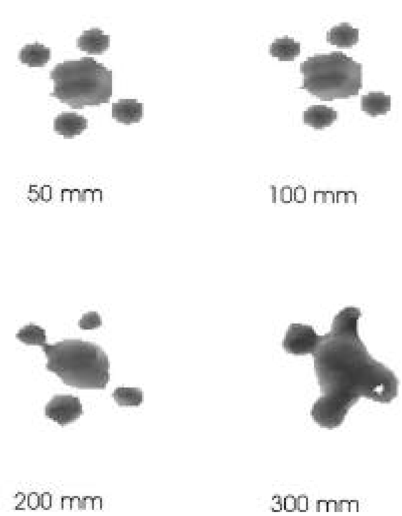

Fig. 4. Simulations of distribution of ^{94}Tc at 4 distances from the source of radioactivity. The source was a 4 mm cube with four 2 mm cubes located four corners. Detector energy and position resolutions were 2 keV and 0.5 mm respectively.

these gamma ray energies. We plan more detailed simulations in the future.

V. Plans

We also plan to conduct a laboratory demonstration of Compton imaging in the near future. Ideally, large arrays of position-sensitive detectors are desirable to undertake a realisitc demonstration of the Compton imaging technique. However, such arrays are not available. It should be possible to undertake such a demonstration with very few detectors, however. Figure 5 shows a configuration where this concept can be demonstrated with only 2 germanium strip detectors. Using a ^{22}Na source placed between the detectors, the locations of the two 511 keV interactions define a pencil beam along which the decay occurred. The Compton cone of the third gamma ray, with an energy of 1275 keV, is used to limit the position of the radioactive decay to a small region. Such a configuration will not have very high triple coincidence efficiency, but should provide a good demonstration of the coincidence Compton imaging technique.

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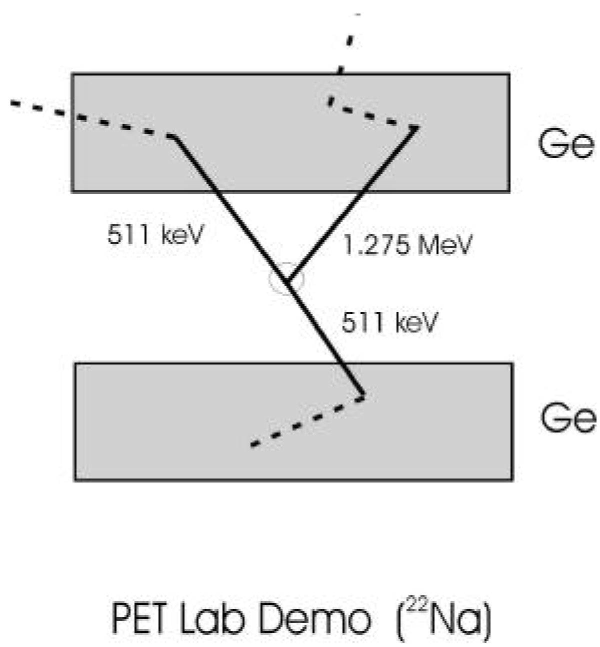

Fig. 5. Planned laboratory demonstration of coincident Compton imager using two position-sensitive germanium detectors and a ^{22}Na source.